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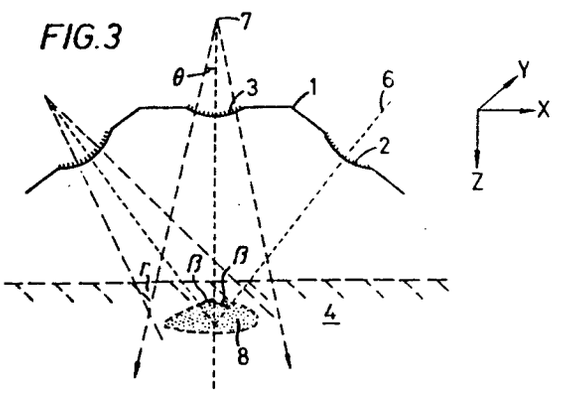
(54) Divergent ultrasound arrays.

(57) An ultrasound hyperthermia unit for the treatment of a target tissue, includes an ultrasound transducer array (1) angled to direct sonic energy toward a target (8), wherein the array (1) comprises a plurality of divergent transducers (2) arranged radially of a central axis to provide a coherent treatment field about a central axis.

Temperature sensing means (10) maybe operative associated with the array and adapted to provide an output signal indicative of temperature values adjacent a target tumour, said signal being utilized to control the output of the transducers.

A liquid bath (5) may be provided for location between the array (1) and the skin (4) of a patient, said liquid bath being adapted to warm or induce local hyperthermia in the skin surface. The divergent array, when associated with a water bath allows a particularly effective treatment of physiotherapeutic conditions or neoplastic tissues without previously experienced colateral damage.

FIG.3



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"DIVERGENT ULTRASOUND ARRAYS"

DESCRIPTION

The present invention relates to divergent ultrasound arrays of transducers adapted to treat medical conditions responsive to ultrasound such as physiotherapeutic conditions and the treatment of neoplastic tissues by ultrasound hyperthermia.

In physiotherapy it has been known to use an ultrasonic transducer in contact with the skin to treat various pain inducing conditions in the joints for example. The use of these ultrasound transducers of the convergent variety is limited because the transducer head must be in intimate contact with the skin, and in that the power output cannot be raised too far because local heating of the skin exceeds patient tolerence. Accordingly, only fairly low ultrasound intensities can be applied to affected parts; ultrasound treatments in depth are not possible.

Focused convergent arrays are also impractical because of the difficulties in providing an acoustic focus at an appropriate point having regard to the presence of bone etc., particularly with the added requirement that with most therapeutic conditions no invasive temperature sensors can be utilized.

Local hyperthermia, in which temperatures normally in the range between 42° and 46°C are maintained for periods from a few minutes up to several hours in neoplastic tissue, has been found to be beneficial for the treatment of cancer in cases where the whole tumour can be properly heated and the surrounding normal tissues can be spared. Many methods of producing local hyperthermia in cancer patients have been suggested and tested. Recently treatment of superficial human neoplasms by local hyperthermia induced by ultrasound has been shown to be useful.

For ultrasonically-induced local hyperthermia the elevation of temperature inside tissue is affected by the ultrasonic absorption, the blood perfusion rate, the thermal conductivity of the tissue, and the choice of transducer parameters. It has been shown that the

temperature distribution produced by a single convergent focused ultrasonic beam is very sharp even in the steady-state situations where the thermal conduction and blood perfusion smooth it. Moreover, we have found that the treated volumes are usually not large enough for most of the clinical cases.

In our European Application No: 0 111 386 we have described an array of sonic transducers radially disposed about an axis, angled to direct sonic energy toward said axis at a point of acoustic focus remote therefrom, and temperature sensing means for association with the point of acoustic focus whereby a or each transducer is adapted for relative angular movement thereby to allow adjustment of the shape and/or position of the acoustic focus in response to an output from the temperature sensing means.

A system utilizing circulating warm water in conjunction with multiple divergent ultrasonic beams to achieve local hyperthermia has also been described in the above European Patent Application.

When a single steered, focussed ultrasonic beam is used, only a small portion of the tumour tissue is heated at one time, the elevation of temperature inside tumour is completely dependent upon the averaging effect induced by thermal conduction and blood perfusion in tissue. The possibility of under-heating or over-heating a certain portion of tissue exists as the blood perfusion rate will change during treatment because of vasodilation. Furthermore, lack of homogeneity in blood flow and inhomogeneity in ultrasonic absorption in tumour will greatly affect the temperature distribution. Moreover, a single steered and focussed ultrasonic beam takes a longer time to heat up a whole tumour to a therapeutic temperature level, and also involves much higher instantaneous intensities.

The array in accord with our European Patent Application No: 0 111 386 although effective requires in practice the provision of computer controlled stepping motors to alter the focal points of the movable transducer array in response to outputs from said temperature sensing means, and this adds to the complexity and cost of the device.

We have now found that excellent therapeutic results can be achieved by using a divergent ultrasonic field in conjunction with warm water circulating over the skin above the treatment area to achieve local hyperthermia. This has some advantages because water circulating at 43°C raises local tissue temperature at a depth of 3cm below the skin to 40°C. Thus the intensity of ultrasound required to produce therapeutic temperatures can be much reduced, with the benefit of minimizing any potential side-effects of the ultrasound. Moreover, divergent fields are able to cover a bigger treatment volume and spare the organs beneath the treated area because of the reductions in local intensity. Further the microprocessor/microcomputer control is relatively simple since it does not need to control stepping motors.

According to the present invention therefore there is provided an ultrasound hyperthermia unit including an ultrasound transducer assembly angled to direct sonic energy toward a target,

characterised in that the array comprises a plurality of divergent transducers each directed to

provide a treatment field about a central axis.

A temperature sensing means for operative association with the array, and adapted to provide an output signal indicative of temperature values, adjacent a target tumour may be provided when the unit is adapted for use in the treatment of neoplastic tissues. In this instance the arrangement is such that said output signal is utilised to control the power output of the array to achieve localised heating of the tumour tissues above viability.

In both the above foregoing instances (physiotherapeutic treatment and the treatment of neoplastic tissues) a liquid bath may be located between the array and the skin of a patient, said liquid bath being adapted to warm the superficial tissues. Optionally, the assembly may be provided with a transducer coaxial with the central support axis, and with a plurality of divergent transducers arranged radially about the central axis.

A system in accordance with this invention may comprise a warm-water circulating unit for contact with the skin of a patient, which forms a single unit with a transducer array, of preferably four or five divergent transducers. The bath may be formed in part of a flexible membrane so that it can change its shape in use to modify the distance between the array and the superficial tissues. One transducer may be a plane transducer and may be placed centrally with the other four or five transducers symmetrically placed around it, but o ^ inclined to it at an angle of, for example, 66.4 to the main axis of the central transducer.

In a preferred embodiment the tranducer array is formed of a support into which selected tranducers may be plugged as necessary. The assembly of the invention may also be provided with a liquid lens in association with each transducer to alter the ultrasound distribution for different treatments.

In a preferred embodiment the transducers may be interspersed with one or more imaging or temperature sensing transducers to allow the array to provide further information. One of the ultrasound

transducers can, for example, also be used intermittently in a sensing role, or the additional transducer can be used in a pulsed or phased sensing mode.

The temperature sensing means may be a thermocouple or other sensing device adapted for positioning adjacent or in the tumour tissue. This is preferably a linearly arranged sensor with a plurality of sensing positions along its length whereby a single needle insertion will provide all the required temperature information for microprocessor control.

Alternatively, non-invasive probes may be used such as microwave thermometers.

Additionally, the skin overlaid by the bath may have in contact therewith a liquid crystal device to indicate the temperature of the skin by colour changes.

The assembly of the invention may be controlled by a microprocessor which utilizes the temperature signals to control the power output thereof. This enables field contours, (ie: isotherms) to be arranged to conform with non-symmetric tumour shapes. Further by

this or other means the frequency of the ultrasound may be stepped up or down to harmonic, higher harmonic or sub-harmonic frequencies to alter depth, temperature distributions since we have found that adsorption coefficient depends on frequency. Further square wave or sinusoidal signals may be selected to alter temperature distribution.

Simpler assemblies are usually required for use in physiotherapy. Generally, the temperature sensing meas can be dispensed with, and the unit will operate in without a water bath so long as the array is in intimate contact with the skin. However a water bath enables the unit to be readily used on difficult body contours such as elbows and knees, and gentle warming of the skin in these areas assists treatment by reducing the amount of ultrasound required.

Because ultrasound is directed toward a source of pain from, for example, five different directions, higher intensities can be applied to a target joint, for example, without affecting patient tolerance. The use of divergent transducers allows large volumes of tissues to be treated without adversely affecting

underlying sensitive tissues because the applied power decreases exponentially.

The invention will now be described by way of illustration only with reference to the accompanying drawings which show:-

Figure 1:
The top-view of a five transducers array for use in physiotherapy or the treatment of neoplastic tissues.

Figure 2:
The side-view of the five transducers array of Figure 1 in conjunction with a water bath.

Figure 3:
The geometry of the transducers array of Figures 1 and 2 in use.

Figure 4:
The iso(pressure)^{2}curves with the increment of 12.5% of the peak value in an axial plane for the five divergent transducers array in water (Frequency =1.1 MHz, diameter of transducer = 5 cm, radius of

curvature = 12.1 cm and the voltage fed into the central transducer is five times that fed into the other transducers).

Figure 5a.

The iso(pressure)^2 curves with the increment of 12.5% of the peak value in a transverse plane at z=15.6 cm for the five divergent transducers array in water (Same conditions as Figure 4).

Figure 5b:

The field pattern registrated on a piece of paper which was placed 15.6 cm away from the front edge of the array (The method of experiment is described in the text.)

Figure 5c:

The pressure-squared distribution along the x-axis in Figure (5b) (All parameters are the same as that in Figure (5a) except that the interval between each data point is about 1/3 wavelength of the ultrasound beams).

Figure 6a:

The iso(pressure)^2 curves with the increment of 12.5% of the peak value in a transverse plane at z=5.6 cm for the five transducers array in water (Same conditions as Figure 4).

Figure 6b:

The field pattern registrated on a piece of paper which was placed 5.6 cm away from the front edge of the array.

Figure 6c:

The pressure-squared distribution along the x-axis in Figure (6b).

Figure 7:

The model used in the calculation of temperature distributions: a volume of 10*10*10 cm^3 of tissue has a tumour volume of 3*3*3 cm^3 which is situated at 1 cm beneath the skin.

Figure 8:

The temperature distribution in a longitudinal plane calculated from the 3-D simulation model.

Figure 9:

The temperature distribution in the transverse plane (2 cm deep from the skin) calculated from the simulation model.

Figure 10:

The temperature distribution in the transverse plane (2 cm deep from the skin) calculated from the simulation model.

Figure 10:

The temperature distribution in the transverse plane (3 cm deep from the skin) calculated from the simulation model.

Figure 11:

The temperature distribution in the transverse plane (4 cm deep from the skin) calculated from the simulation model.

Figure 12:

The temperature distribution in the transverse plane (5cm deep from the skin) calculated from the simulation model

Figure 13:

The axial temperature distribution calculated from our computer model versus depth after 2.5 minutes of sonication under the assumption that the blood perfusion rate is 10% (□) and 50% (◇) of the peripheral respectively.

Figure 14:

The temperature distribution in a longitudinal plane calculated from the simulation model under the assumption that the blood flow is normal tissue is 9.6 ml/Kg/sec. (The other conditions are the same as that in Figure 8.)

The above figures show the theoretical calculation of the field patterns produced by an array of transducers comprising four equally spaced transducers disposed about a central transducer. The equally spaced transducers are arrayed at a fixed angle to the central axis. A computer simulation model to simulate the effects of blood flow, thermal conduction, ultrasonic absorption in tissues, skin temperature and different power setting of the transducers on the temperature distribution inside tissue during local hyperthermia is also shown. The method is applicable

to the predication of temperature distributions in tissue being different numbers of transducers and different geometrical configurations.

With particular reference to Figures 1 to 3 a substantially rectangular transducer array 1 provides a frame for the support of a plurality of radially spaced divergent transducers 2 in equally spaced relation. The angles of the transducers and their configuration are apparent from Figure 2, from which it will be noted that each transducer is a divergent transducer, and that although each transducer is directed toward a common focal point when the central axis of each transducer are considered, each is capable only of providing a divergent sonic field, over a certain predetermined area wherein a target lies. This target may be a knee joint or a neoplastic tissue.

The central transducer 3 is provided about the central axis of the array and may, if desired, be adapted to perform special functions.

With reference to Figure 2, the array and water bath assembly, in accordance with the present invention is shown in strictly diagrammatic form. Water entering the water bath 5 from the left hand side of the Figure 2 exits from the water bath and transducer assembly to the right hand side of the Figure 2 as shown in the drawings. The water maintains the skin temperature at a desired temperature, for example when treating neoplastic tissues about 42 C^{o} in the absence of any power from the transducers. The skin 4 overlying a target for example a tumour shown diagrammatically at 8 has a plurality of sensors 10 inserted therein in juxtaposition with, and into the tumour 8. Outputs from the sensors 10 are fed to a computer which can be utilized to control the power of each transducer separately. The sensers 10 are not required for physiotherapy since the power output can be controlled on an empirical basis.

The transducer array is operatively connected to the water bath such that the induced sound passes through the water bath into the patient with a minimum of power loss. This ensures that minimum levels of

ultrasonic power is required to achieve maximum heating of the target area.

As is shown in Figure 3 the central axis 7 provided to the central transducer, whereas the peripheral axis 6 indicates a point of conjunction of the axis of adjacent transducers. This is not, however, an acoustic focus since the acoustic transducers are of the divergent type.

The unit of Figures 1 to 3 may be adapted for use physiotherapeutically in the following conditions:-

Trauma to tibia
Painful arc syndrome
Medial meniscus lesion
Degenerative changes
Pain - thoracic vertebrae
Tenderness over rhomboid
Bruised perineum and haemorrhoids
Haematoma of line of episiotomy
Bruised oedematous perineum
Rotator cuff lesion
Contraceptive and induration of delto-pectoral flap
Torn attachment - abdominal muscles (sports injury)

Medical ligament - knee

Chronic strain of abdominal muscles

Haematoma soleus

Plated ankle

Biceps femoris strain

Bicepital haematoma

Acute tear ligament; med. ligament (knee)

Pain paraesthesia distribution T10

Laceration dorsum of hand

Sacro-iliac strain

Ligament injury - ankle

Traumatic injuries: foot, thumb

Dislocation of shoulder

Contusion - shoulder joint (MS)

Sprain -ankle

Pre-tibial inflammation

Muscle contusion with alcoholic neurities overlay

Injury to acromio clavicular joint

Back strain

Swelling and bruising: tibia, ankle

Effusion and bruising of knee

Herpes Zoster

Haematoma above L. lat. malleolus

Undisplaced fracture -styloid-process radius

Soft tissue injury

Effusion R. pre-patellar bursae

Acute torticollis

Painful Trapezius

Strain-Trapezius and Rhomboids

Epicondylitis

Vastus muscle strain

Pain following Colles fracture

Strained hamstrings

Tendon repair: thumb

Divided flexor tendons

Tenosynovitis extensors: wrist

Gastrocnemius

Referred arm pain

Adhesions following road traffic accident-knee

Chondromalacia - knee

Adherent scar - (swollen fingers)

Theory

(A) Calculation of the field pattern produced by the array of five divergent transducers

In linear acoustics the sound waves caused by the radial motion of a sphere are governed by the first order wave equation

$$\nabla^2 p + k^2 p = 0 \qquad \text{Eq(1)}$$

when we consider only harmonic time variations for the acoustic pressure p (where k=w/c, w is the angular frequency, c the sound speed). If we take a particular case of a polar cap whose surface vibrates with a velocity $u_o \exp(-iwt)$, the half-angle subtended by the curved surface at the centre of curvature is θ_o, and the radius of curvature of the polar cap is b, then the solution of the differential equation (Eq1) can be expressed as

$$p(r,\theta,t) = \frac{\rho c (1-\cos\theta_o) u_o \exp(-iwt)}{4i} \sum_{n=0}^{\infty} \frac{P_{n-1}(\cos\theta_o) - P_{n+1}(\cos\theta_o)}{h_n'(kb)} h_n(kr) P_n(\cos\theta) \qquad \text{Eq(2)}$$

where P is the density of the medium in which ultrasound is propagating, P stands for Legendre function, h stands for spherical Hankel function and the prime in the denominator indicates differentiation with respect to the argument kb. This series has been shown to be uniformly convergent in any region r>b (Wilcox 1956).

The ultrasonic field produced by a divergent transducer shaped like a polar cap can be described by Eq(2). When an array of five transducers is

considered, the resultant pressure at the point (r,0) is the vector sum of individual pressure at that point produced by each transducer. Although the particle velocity near a vibrating surface is not in phase with the pressure, the particle velocity will be in the radial direction and in phase with the pressure when it is far from the vibrating surface (Morse etal, 1968). Hence the resultant intensity will be proportional to the square of the resultant pressure when kr is very large. Mathematically, it is expressed as follows:

$$I \propto [p(\text{total})]^2 = \left| \sum_{i=1}^{5} p_i \right|^2 \qquad \text{Eq(3)}$$

A computational program has been developed to calculate the intensity distribution generated by the array of five divergent transducers where each transducer is driven with the same sinusoidal frequency but with the different vaoltage.

(B) Computer simulation model of temperature distribution in tissue during local hyperthermia induced by our five transducers array

The general heat conduction equation

$$\rho S \frac{\partial T}{\partial t} - k\left(\frac{\partial^2 T}{\partial x^2} + \frac{\partial^2 T}{\partial y^2} + \frac{\partial^2 T}{\partial z^2}\right) = H(x,y,z) - C(x,y,z,T) \quad \text{Eq(4)}$$

has previously been employed to simulate the temperature distribution in tissue during local hyperthermia induced by ultrasound (Hynynen etal 1981) Here P is the density of the tissue, S the specific heat, T tissue temperature, t time, k the thermal conductivity of the tissue, xyz space coordinates, H the rate at which heat is being generated by ultrasound in the element (x,y,z) per unit volume and C the cooling function induced by blood perfusion in tissues.

Many standard methods have been employed to solve the general heat conduction equation. Finite-difference approximation is one of the numerical approaches (e.g: Chapman 1974; Hildebrand 1968). For explicit finite-difference formulation, Eq(3) can be transformed into

$$T(x,y,z,t+1)=\lambda[T(x-1,y,z,t)+T(x+1,y,z,t)$$

$$+T(x,y-1,z,t)+T(x,y+1,z,t)\ +T(x,y,z-1,t)+T(x,y,z+1,t)$$

$$-6T(x,y,z,t)]\ +T(x,y,z,t)\ +(\Delta t)/(\rho S)[H(x,y,z)-C(x,y,z,T)]$$

Eq(5)

under the assumption that the spatial increments ($\Delta x, \Delta y, \Delta z$) are equal and λ represents $k(\Delta t)/(\rho S \Delta x^2)$ which is a dimensionless parameter. In order to satisfy stability and convergence criteria, λ must be taken to have any positive value less than 1/6 (Chapman 1974).

It is accepted that the blood perfusion rate in most tumours is usually lower than that in the surrounding normal tissues, but the blood perfusion rate at the advancing front (tumour periphery) is often much higher. Therefore, we have developed a computer simulation program to simulate the effect of perfusion on temperature distribution during hyperthermia treatment generated by our 5-element transducer array. Also we take account of the thermal conduction and ultrasonic absorption.

Computer predications

(A) intensity distribution in water

In the present invention there are five transducers in the array. The radius of curvature of each transducer is 12.1 cm and the half-angle subtended by the curved surface at the centre of curvature is 11.9°. The main axis of each transducer is inclined 23.6° with the main axis of the central transducer (Figure 3). Each transducer in the array can be fed with a different voltage in order to achieve different intensity distributions and hence correspondingly different

temperature distributions in tissue. Empirically, in the first instance, the voltage fed into the central transducer is five times higher than that fed into the four outer transducers. The (acoustic pressure amplitude)^{2}distribution in a longitudinal plane (XZ-plane) is calculated by using Eq (2) & (3) and the result is presented in Figure 4. In addition, the (acoustic pressure amplitude)^{2}distributions in the transverse plane at 5.6 cm and 15.6 cm away from the edge of the transducer unit are presented in Figure (5a) and Figure (6a). These show that the acoustic intensity dimenishes rapidly with distance along the axis and varies symmetrically about the XZ- and YZ-planes. The acoustic pressure-amplitude-squared distributions also show that use can be made of the intensity distribution between z=10 and 20 cm for hyperthermia because in that region a large tissue volume will be heated with reasonable uniformity. Moreover, the surrounding tissue and the organs beneath the treated area will be spared because the ultrasonic intensity diminishes very rapidly with distance.

The ultrasonic field patterns were visualized as follows:

the transducer array was immersed in a tank of water with a few drops of methylene blue dye added. Pieces of white paper were sequentially placed at different positions in the path of the ultrasonic beam. The dye particles move under the influence of the ultrasonic field and are trapped in the paper, so that a coloured pattern corresponding to the field distribution is produced on the paper.

The pattern registrated at 15.2 cm away from the array, obtained in this way is shown in Figure (5b). The shape of the pattern is consistent with the computer simulation. However, Figure (5b) shows up a clear interference pattern which is produced by the interference between the five ultrasound beams. Figure (5a) does not show up this pattern. It is because the data points in the mes for producing figure (5a) are relatively sparse (ie: the separation between each data point is greater than the wavelength of the ultrasound beams), so that the interference pattern does not appear in Figure (5a). The data point separation has to be large to ensure realistic computing time. If the separation between each data

point in the mesh is reduced to a value which is smaller than the wavelength, our simulation model will show the detail of the interference pattern.

In fact, the pressure-amplitude-squared distribution along the x-axis with an interval of 0.04 cm between each data point is plotted in Figure (5c). The interference pattern in Figure (5b) and the result of simulation shown in Figure (5c) are consistent. Similarly, Figure (6c) shows the pressure-amplitude-squared distribution along the x-axis in Figure (6b) which shows the field distribution pattern at 5.6 cm away from the front face of the transducer unit.

(B) Temperature distribution in tissue

The actual elevation of temperature in tissue during local hyperthermia can be affected by many factors; different blood perfusion rate in tumour and in normal tissues, thermal conductivity of tissue and ultrasonic absorption. The overall parameters of thermoregulatory response in the human body are very complicated, but the major factor is the cooling function in local hyperthermia which can be mainly

attributed to the blood flow in tissue. A warm-water bath, which is maintained at a constant temperature, say 42°C. was placed on the patient's skin just above the treated area in order to reduce the ultrasonic power and minimize the heat loss through skin in the course of local hyperthermia treatment of superficial neoplasm.

In the 3-D computational simulation model we consider the following conditions: the temperature distribution in a cube 10*10*10 cm^3 is located at 1 cm below the skin (see Figure 7). The blood perfusion rate inside the tumour is assumed to be only 10% of that in the peripheral tissues. The blood perfusion rate in tumour's periphery and normal tissue is 2.3 ml/kg/sec and 0.5 ml/kg/sec respectively. The thermal conductivity of the tissue is 0.44 W/M/K, ultrasonic attenuation is 8 Np/M and the skin temperature is maintained at 42°C. (by placing a warm-water bath on the skin). The spatial temporal average of the ultrasonic intensity in degassed water at z=13 cm is 3.8 W/sq.cm. The patient's skin is 10 cm away from the central transducer. The results of computer simulation are shown in Figure 8 - 12. Figure 8 shows

the temperature distribution in tissue along a longitudinal plane (XZ-plane) after 2 minutes of sonication. It indicates that the maximum temperature 48.4^{o}C is about 2 cm below the skin and the iso-temperature curve of 42.8^{o}C. covers an ellipsoidal volume with a maximum diameter of about 3 cm and 4 cm long, (ie: about 37.7 cm^{3}). All the tissue at 5 cm below the skin is spared. Figure 9, 10, 11 and 12 show the temperature distribution in the transverse plane (XY-plane) at the depth of 2, 3, 4 and 5 cm below the skin respectively. They further support the view that the overlapping divergent fields produce an acceptable heating pattern during hyperthermia treatment of a superficial neoplasm.

The blood perfusion rate can affect the temperature distribution where the blood perfusion rate in tumour is 10% and 50% of the peripheral blood flow respectively and the blod flow in normal tissue is unchanged, the axial temperature distribution in tissue as a function of depth at different times is plotted in Figure 13. Results show that the maximum temperature will be reduced by about 1^{o}C. after 2.5 minutes of sonication if the blood perfusion rate in

tumour is increased from 10% to 50% of the peripheral blood perfusion rate.

Furthermore, the blood perfusion rate in the surrounding normal tissue can also affect the temperature distribution. Where the blood perfusion rate in tumour is 10% and 50% of the peripheral blood flow respectively and the blood flow in normal tissue is unchanged, the axial temperature distribution in tissue as a function of depth at different times is plotted in Figure 13. Results show that the maximum temperature will be reduced by about $1^{o}C$ after 2.5 minutes of sonication if the blood perfusion rate in tumour is increased from 10% to 50% of the peripheral blood perfusion rate.

Furthermore, the blood perfusion rate in the surrounding normal tissue can also affect the temperature distribution. Where the blood perfusion rate in normal tisse is 9.6 ml/Kg/sec (e.g: in normal liver) and the blood perfusion rate in tumour is still 10% of the peripheral, the temperature distribution along a longitudinal plane is shown in Figure 14. It shows a slightly different temperature distribution as

compared to that in Figure 8. The iso-temperature curve of 42.8°c. still covers about the same volume bt te temperature elevation outside the tumour volume is reduced.

Blood_flow_and_Temperature_Distribution

Both the ^{133}Xe clearance method used for a quantitative measurement of perfusion in superficial tumours and the measurement of scattering of ultrasound by blood flowing in breast cancers, show that most superficial tumours have a considerably lower flood flow than that in the surrounding normal tissue. Frequently, a tumour has a well perfused advancing front and a necrotic centre. This inhomogeneity in blood flow in tumours makes it difficult for a focussed convergent transducer, either fixed in position or moving around the treatment field, to achieve a uniform temperature distribution in the whole tumour simultaneously. Moreover, in practice, it is not easy to concentrate the energy very accurately to the treatment zone and thus inhomogeneous temperature distributions must be accepted. This means that after the treatment some of

the cells in the growing zone may still be viable, so that combination of chemotherapy or radiotherapy with hyperthermia is essential.

From the computer simulation calculation the 5-element hyperthermia unit described here can cover a superficial tumour volume of about 38 cm^3. Larger arrays can cover for 100 to 200 cm^3. If it is assumed that the blood perfusion rate in a tumour is only 10% of that at the periphery, the temperature gradient inside the tumour will be less than 2.7^{0}C. per cm with the temperature maximum 48.4^{o}C. at the necrotic centre after about 2 minutes of sonication. The temperature distribution is fairly uniform. The temperature of the tissue outside the tumour and the tissue at over 5 cm depth from the skin are below the therapeutic temperature level, and thus normal tissues are spared.

Temperature distributions inside tissue will be affected hby the ultrasound intensity distribution. Where the central transducer in the array is fed with a voltage which is five times higher than that fed into the other transducers. The results of the simulation show that the shape of the treatment volume

will be ellipsoidal. However, different treatment patterns will be produced to suit different clinical cases by adusting the ratio of voltages to and powers from each transducer. This is controlled by a temperature sensing element, or elements positioned in or about the tumour tissue.

Accurate measurement of temperature distribution inside the treatment volume is important for successful hyperthermia treatment and requires an array of thermocouples for example to be inserted into the tmour.

Use of Warm-Water Bath

The use of a warm-water bath over the treatment field has several advantages: (1) It maintains a constant surface temperature to minimize any fluctuation caused by physical environment. (2) It reduces the radiative heat loss through the tumour surface. (3) It acts as an ultrasound coupling. (4) It reduces the temperature gradient between the skin and the superficial tumour during local hyperthermia.

Thus, the total ultrasonic power required can be reduced but the therapeutic temperature level can still be maintained. Reducing ultrasonic power is beneficial to the patient because any actual or potential side-effects induced by ultrasound will be minimized. However, the warm-water bath itself cannot achieve a thermpeutic effect even for superficial neoplasms because the heat penetration is not deep enough to treat a neoplasm and the hottest portion is then on the skin. When the divergent-transducer array is used in conjunction with the warm-water bath, it generates a uniform heating pattern suitable for treating superficial tumours lying between 0.5 cm and 4 cm below the skin.

In conclusion, according to the calculations of the ultrasonic fields generated by the five divergent transducers array and the computer simulation model of the temperature distribution inside tumours, the hyperthermia unit of the invention produces a relatively uniform heating pattern and is able to treat bigger tumour volumes and spare the surrounding normal tissues. Thus the unit is particularly suitable for treating superficial tumours lying at between 0.5 cm and 4 cm depth below the skin.

CLAIMS:

1. An ultrasound hyperthermia unit including an ultrasound transducer assembly (1) angled to direct sonic energy toward a target (8),

characterised in that the assembly (1) comprises one or a plurality of divergent transducers (2) arranged to provide a treatment field about a central axis.

2. A unit according to claim 1 characterised by a liquid bath (5) for location between the array (1) and the skin (4) of the patient, said liquid bath being adapted to warm the superficial tissues.

3. A unit according to claim 2 characterised by

temperature sensing means (10) for operative association with the array, and adapted to provide an output signal indicative of temperature values adjacent the target tumour, an output signal being utilized to control the power output of the transducers.

4. A unit according to claim 2 characterized in that the bath is formed in part of a flexible member, and in that the bath can change its shape to modify the distance between the array and the superficial tissues.

5. A unit according to any preceding claim characterised in that the assembly comprises a plurality of divergent transducers (2) arranged radially of the central axis, and has a transducer (3) coaxial with the central axis.

6. A unit according to claim 5 characterised in that said transducers (2.3) are interspersed with one or more imaging or temperature sensing transducers.

7. A unit according to any preceding claim characterized in that a selected transducer (2.3) is utilizable intermittently in a sensing role.

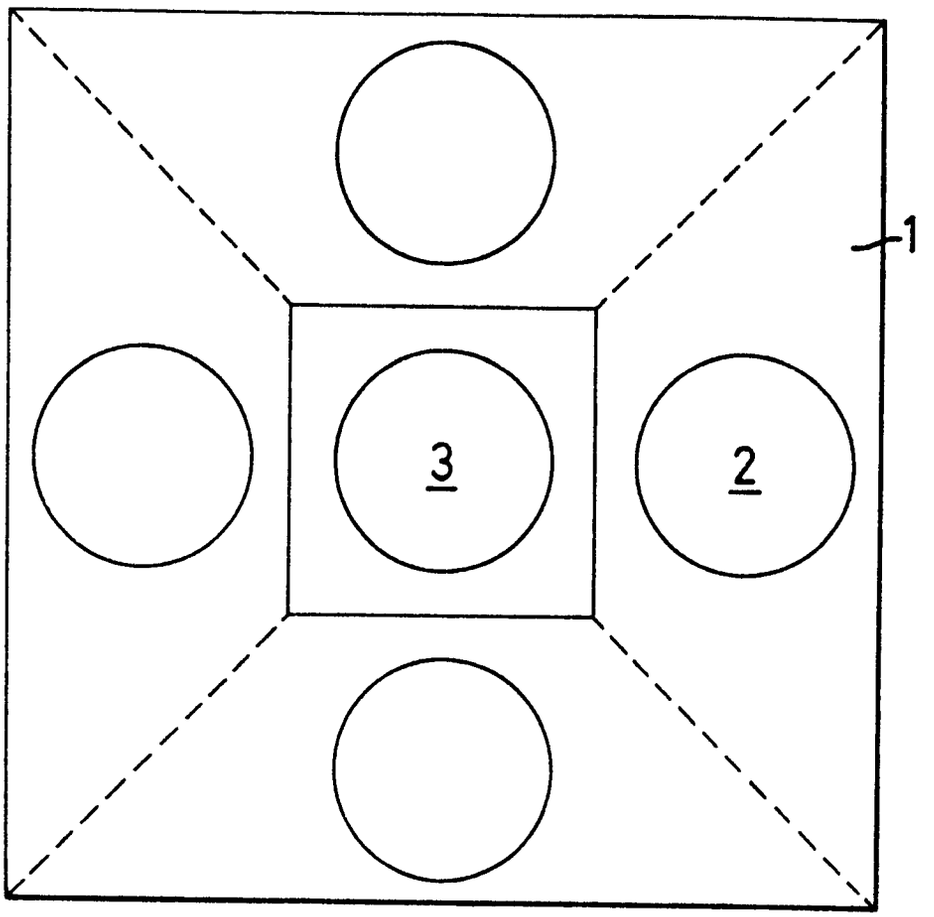
8. A unit according to any preceding claim characterized by means for the control of ultrasound frequency whereby harmonic, higher harmonic or sub-harmonic frequencies can be selected as a control

function to alter temperature depth characteristics of the unit.

9. A unit according to any of the preceding claims characterised in that a liquid lens is associated with each transducer to vary the focal length and focal volume of the unit.

0225120

FIG.1



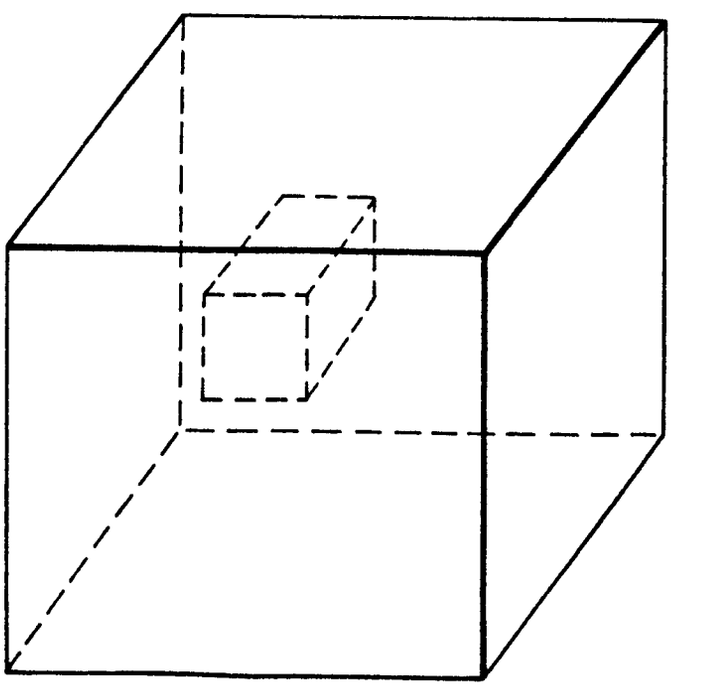


FIG.7

FIG. 2

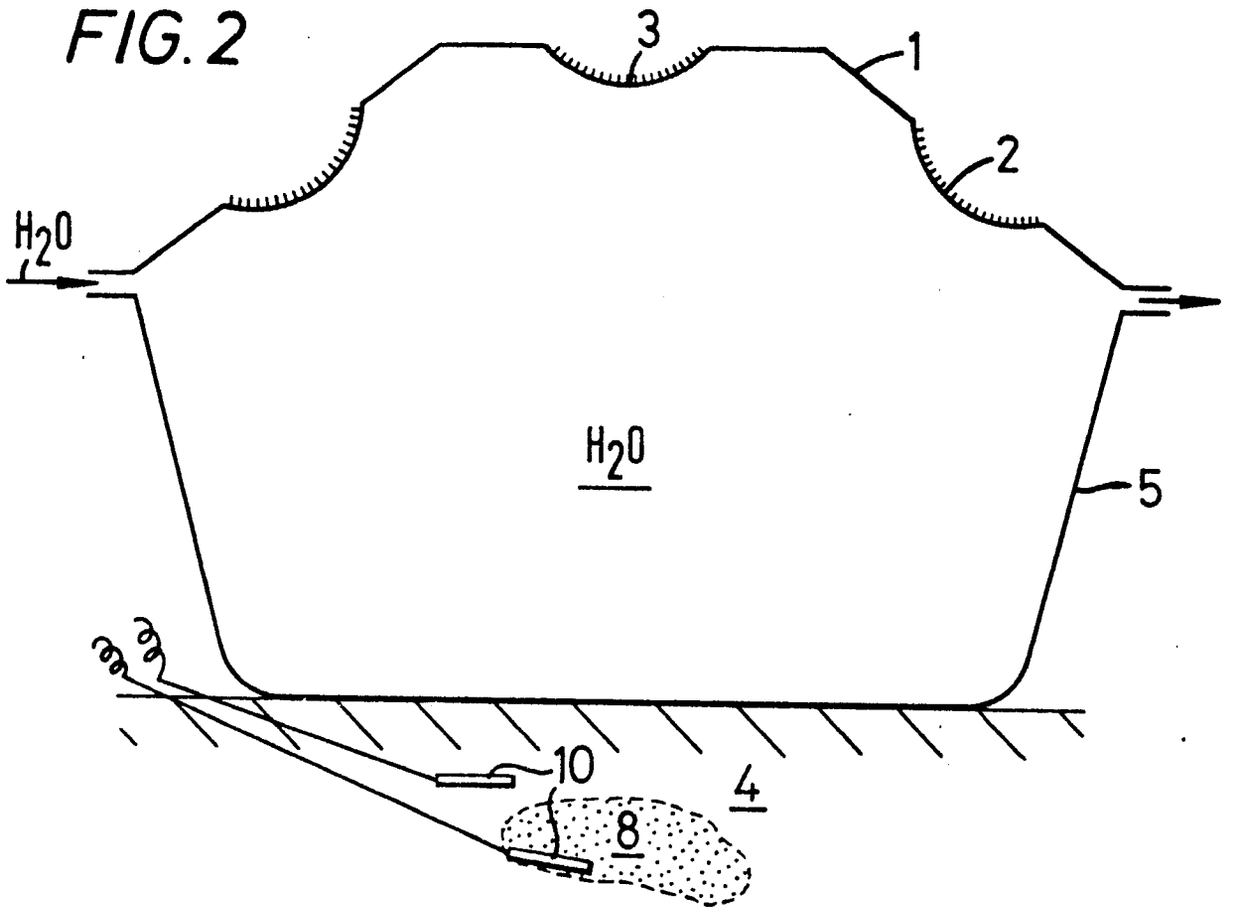


FIG. 3

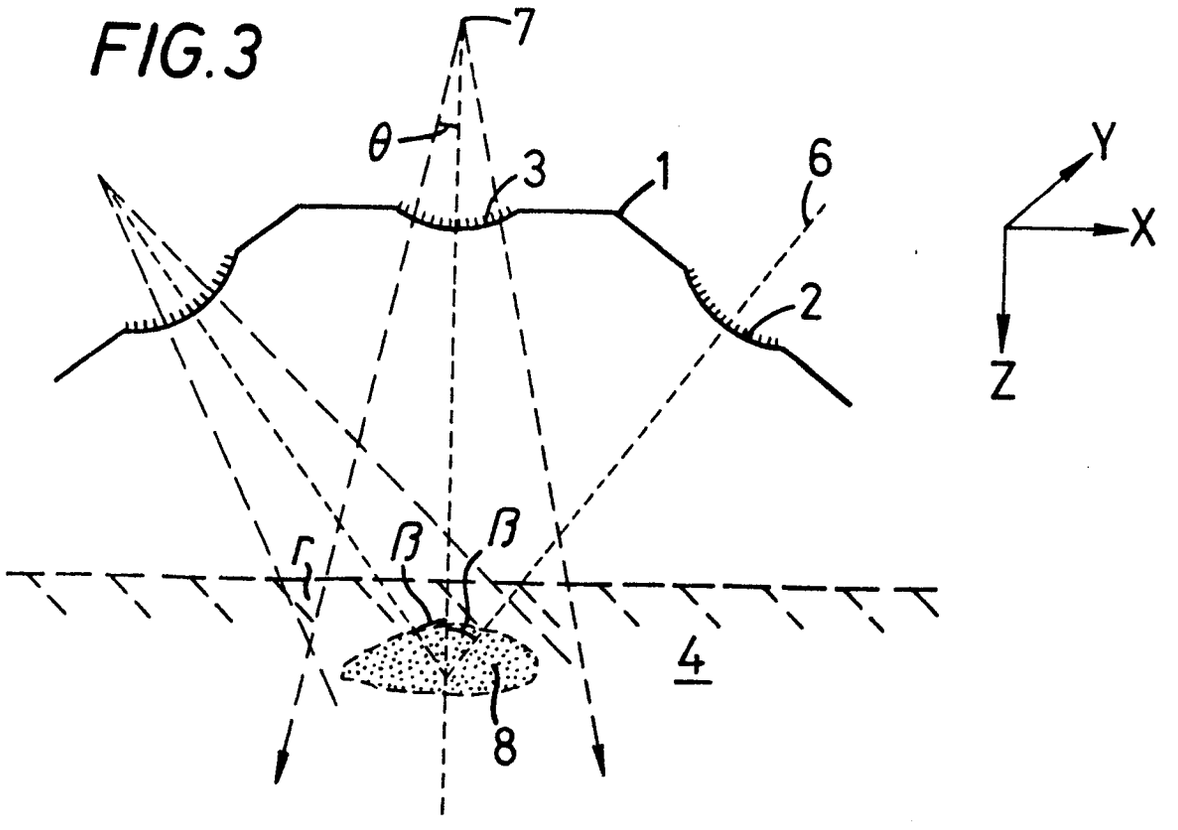
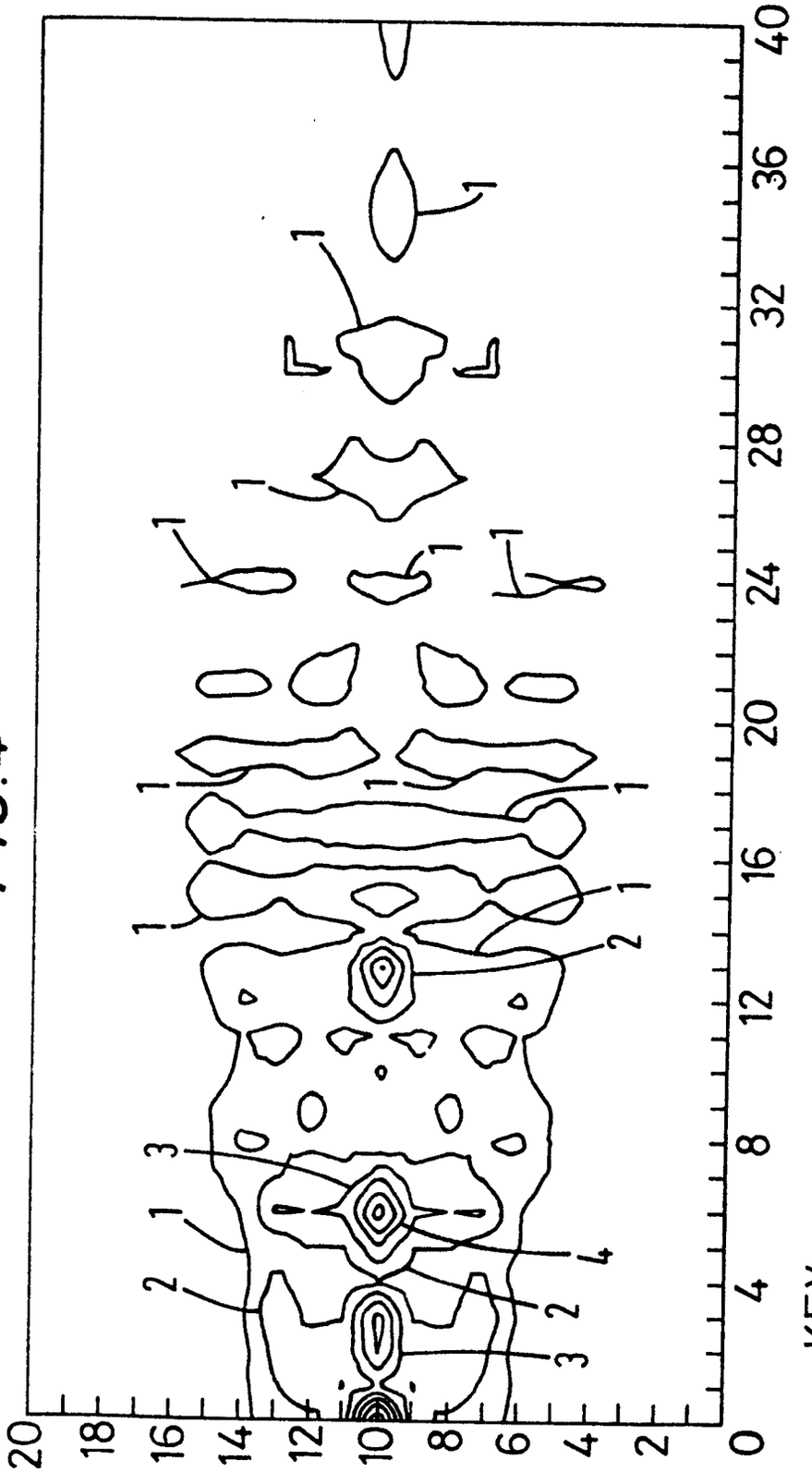


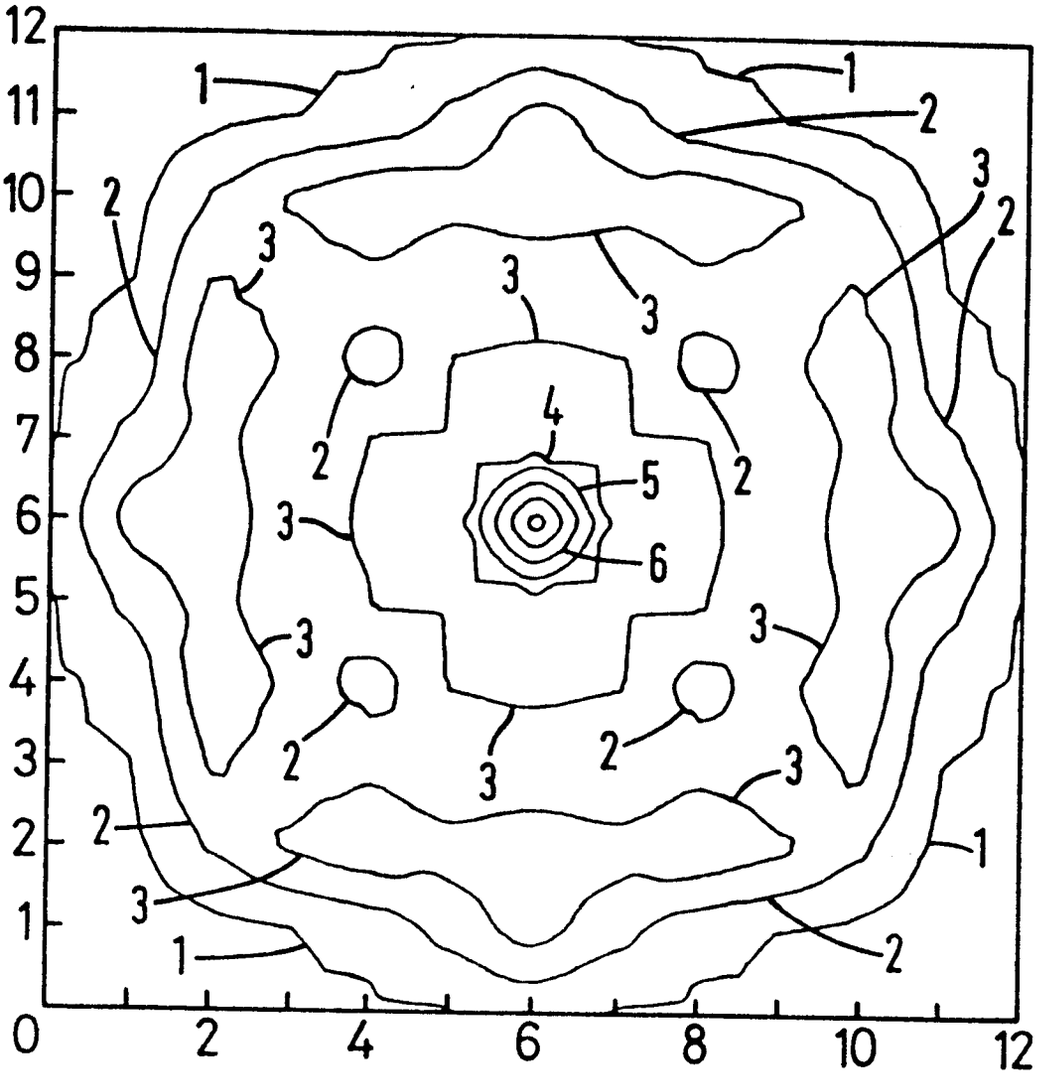
FIG.4



KEY:

CONTOUR NO.	1	2	3	4	5	6	7	8
% OF THE MAXIMUM VALUE IN THAT PLANE	6.25	18.75	31.25	43.75	56.25	68.75	81.25	93.75

FIG.5a



KEY:

CONTOUR NO.	% OF THE MAXIMUM VALUE IN THAT PLAIN
1	6.25
2	18.75
3	31.25
4	43.75
5	56.25
6	68.75
7	81.25
8	93.75

FIG. 5b

FIG.5c

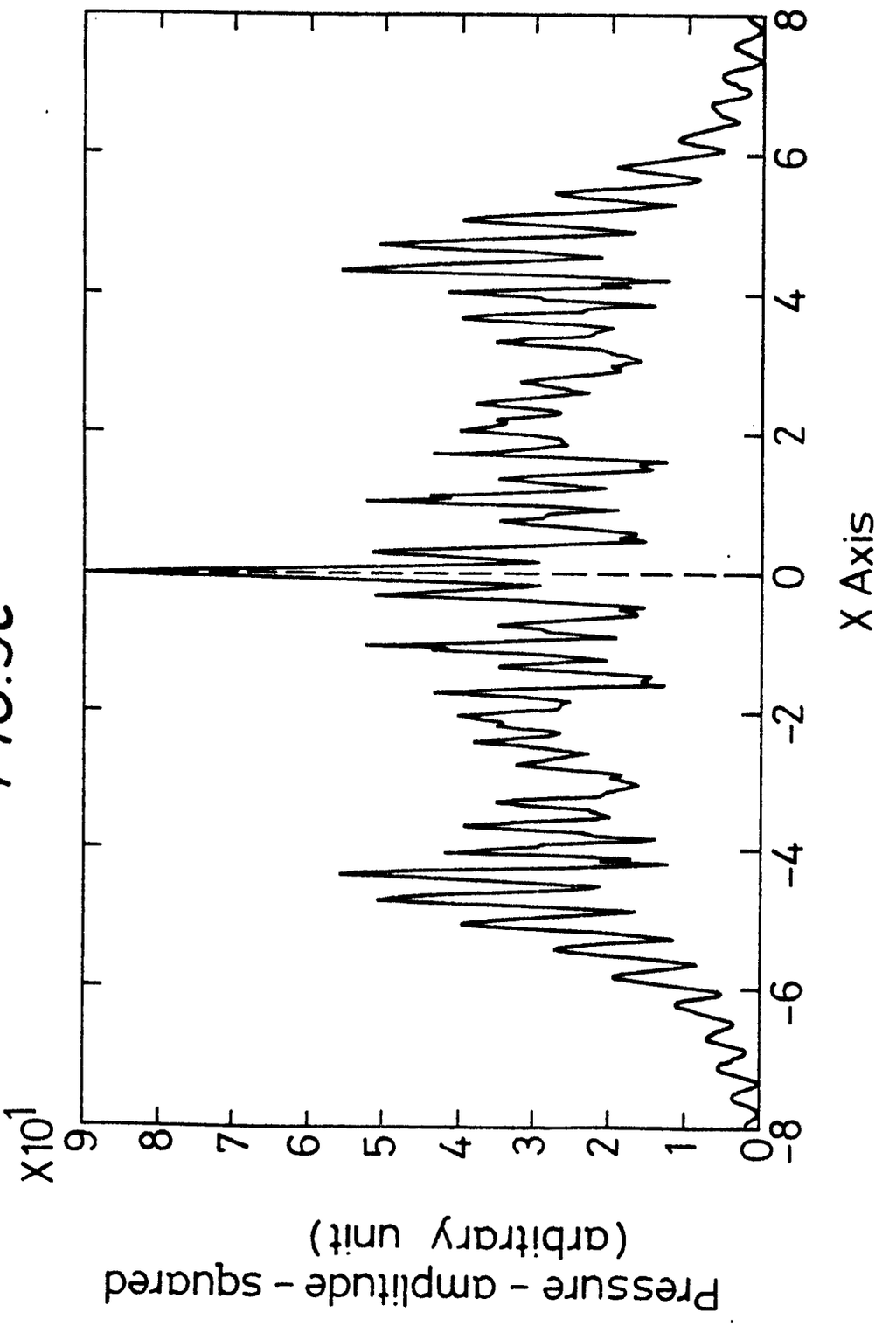
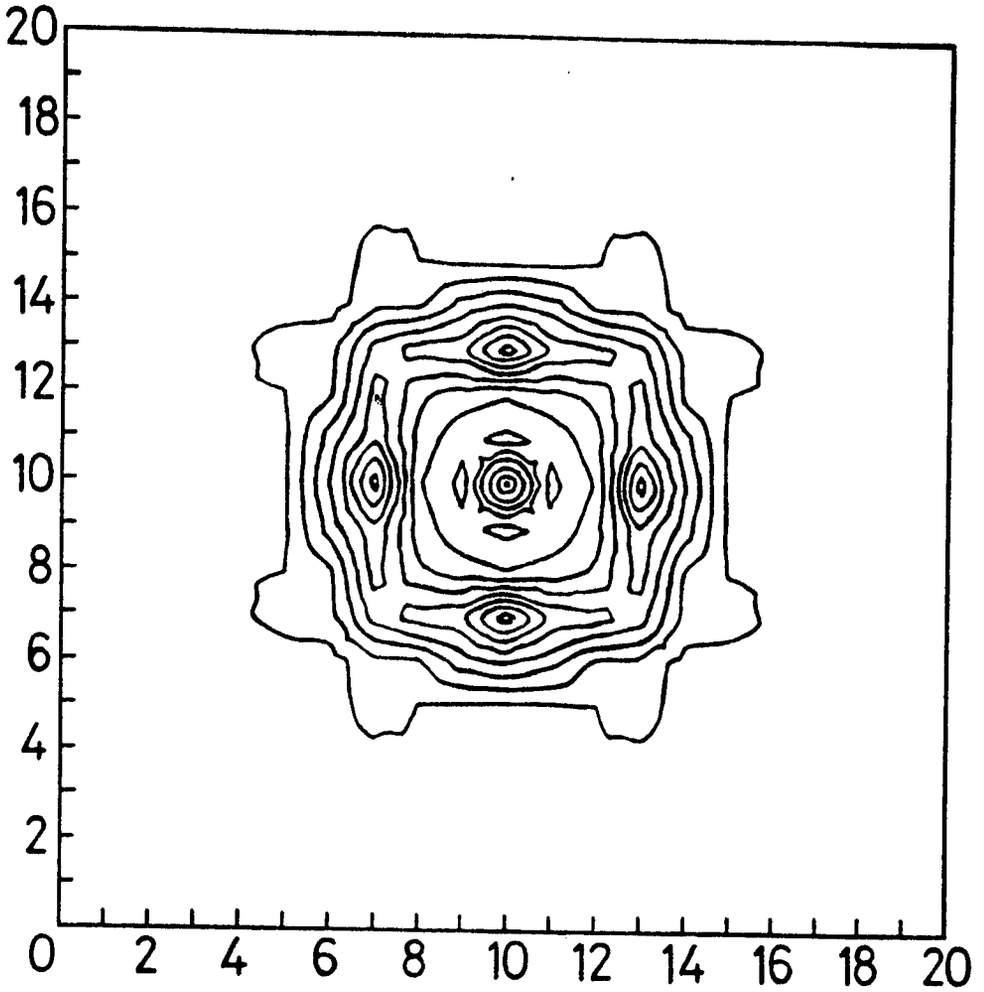

Pressure - amplitude - squared
(arbitrary unit)
X10¹
9
8
7
6
5
4
3
2
1
0
-8
-6
-4
-2
0
2
4
6
8
X Axis

FIG. 6a



KEY:

CONTOUR NO.	% OF THE MAXIMUM VALUE IN THAT PLAIN
1	6.25
2	18.75
3	31.25
4	43.75
5	56.25
6	68.75
7	81.25
8	93.75

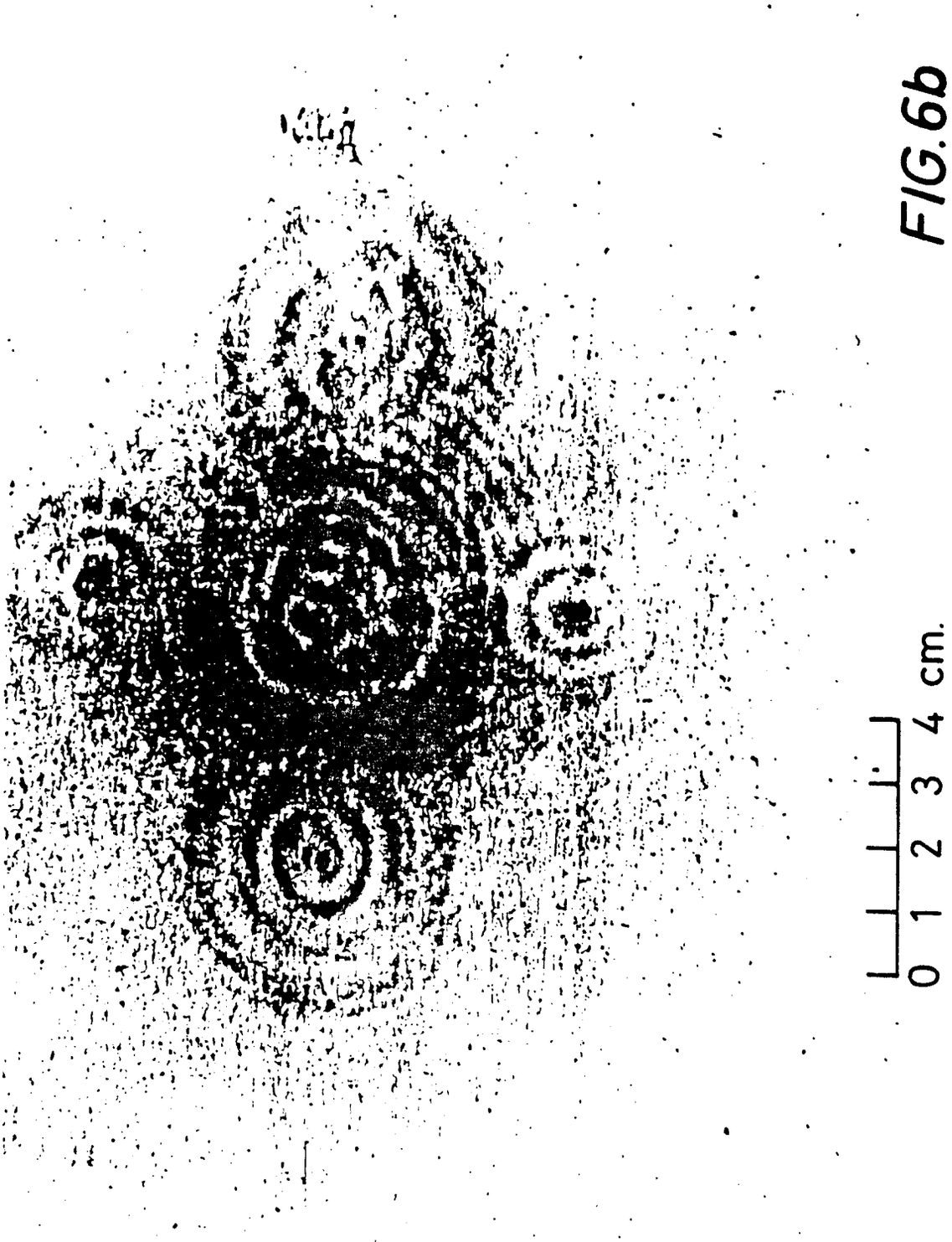


FIG.6b

FIG. 6c

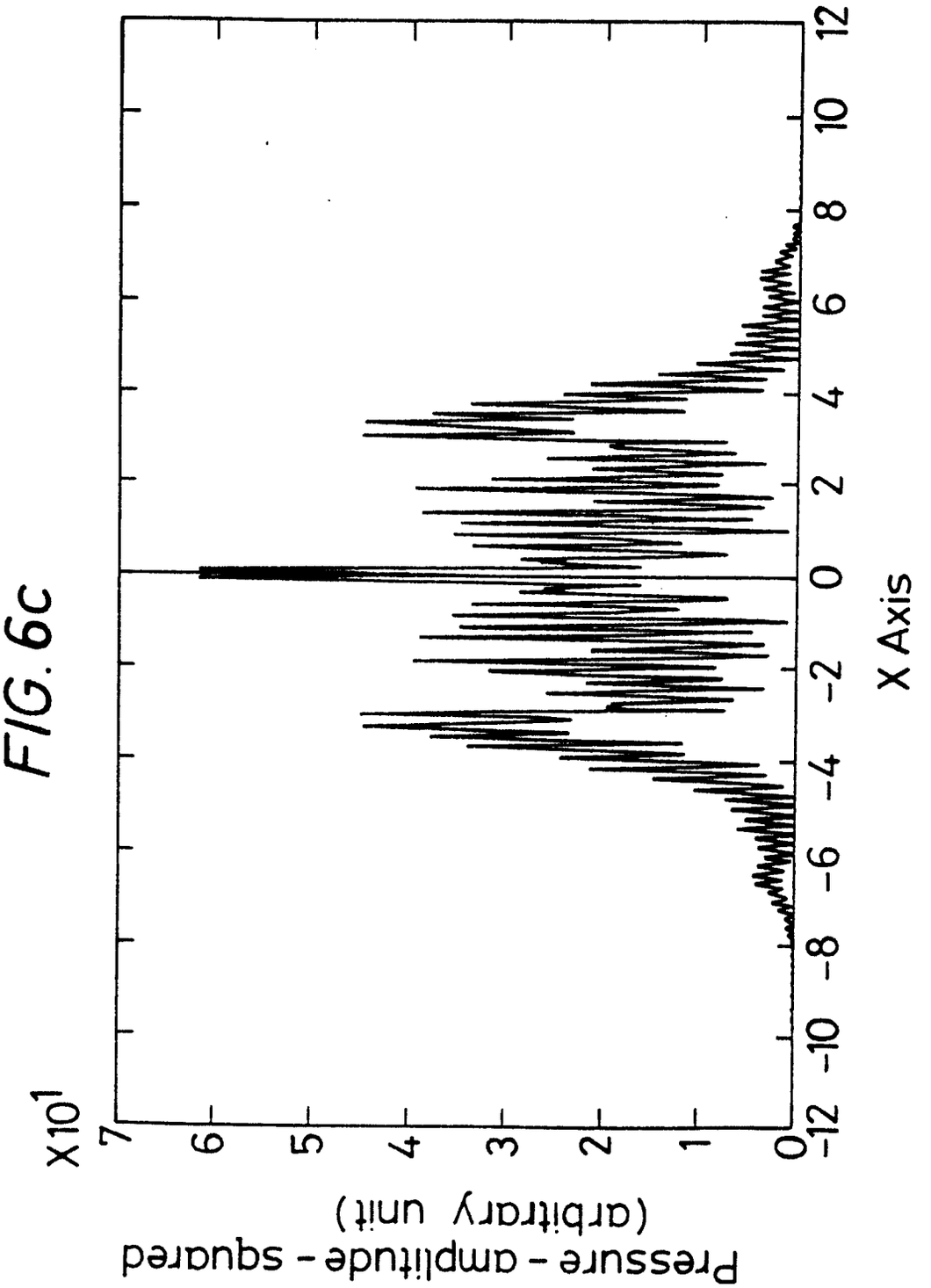
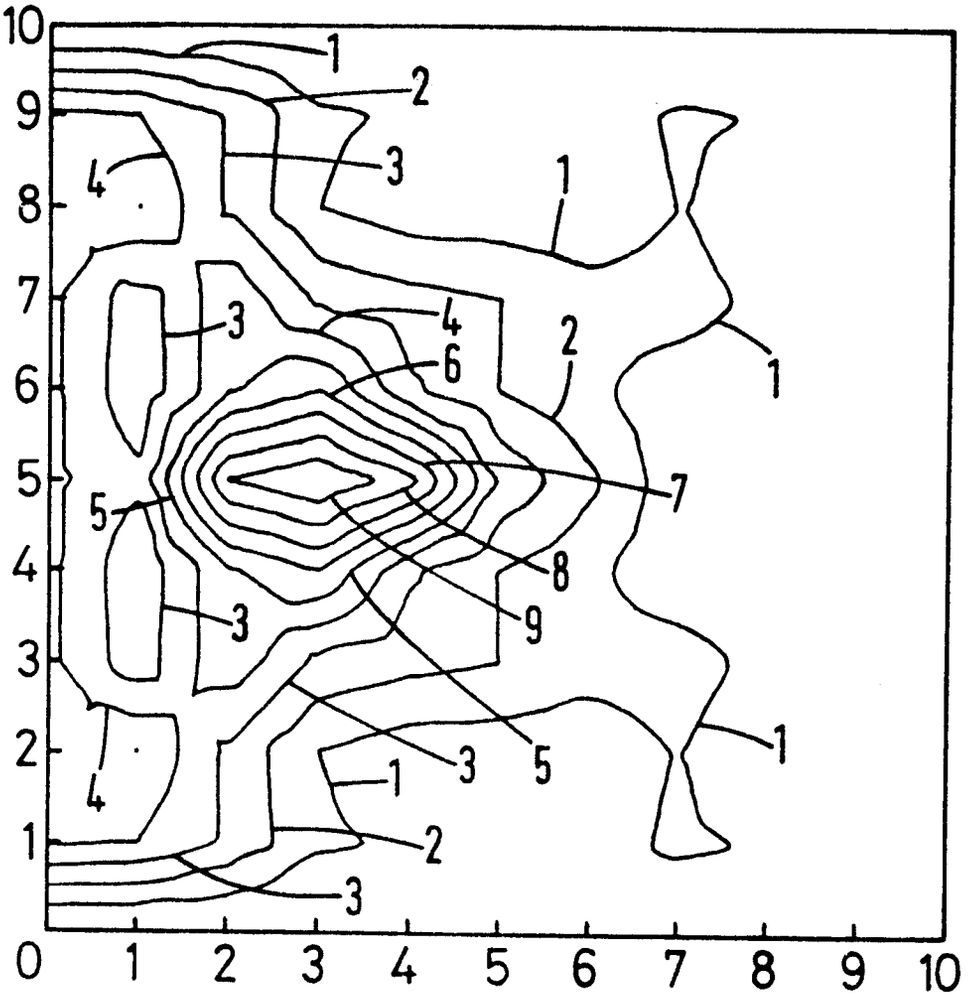
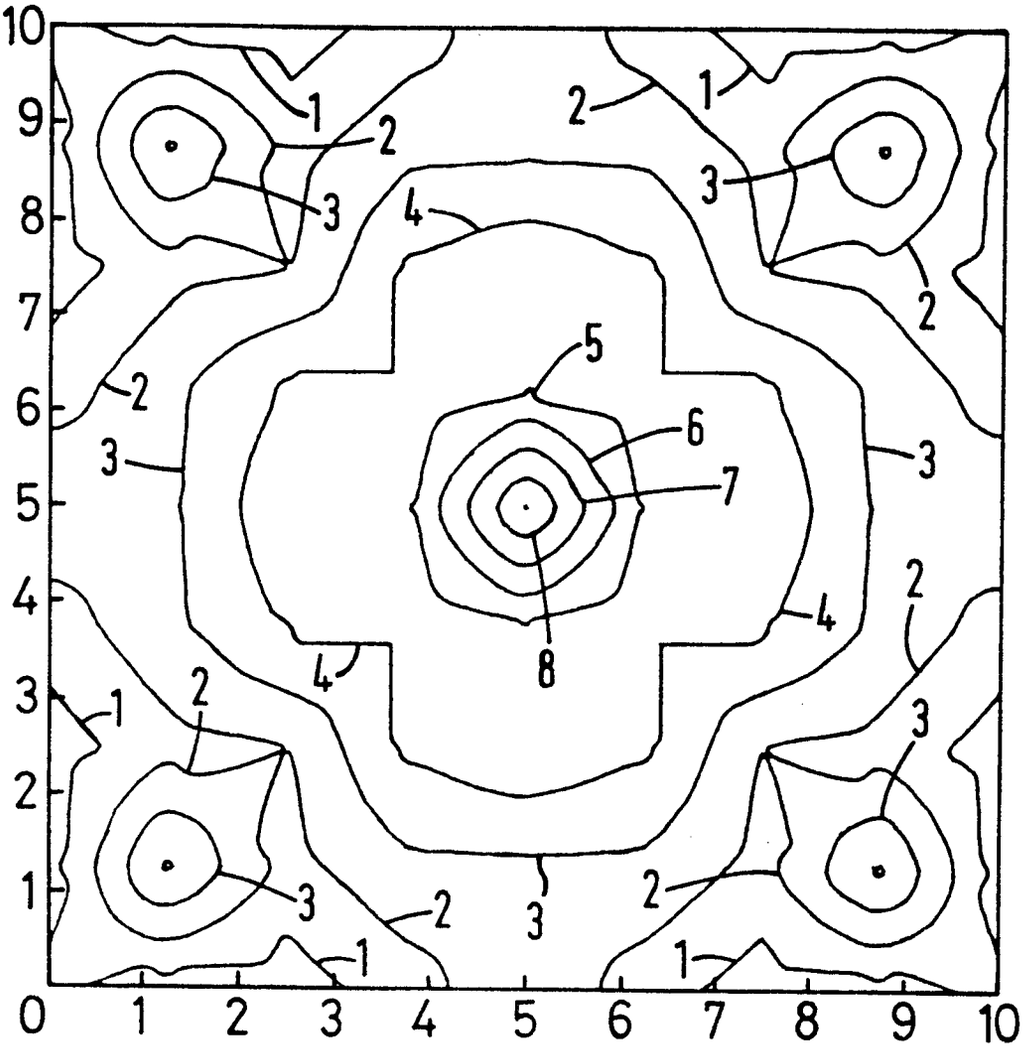

Pressure - amplitude - squared
(arbitrary unit)
X10^1
7
6
5
4
3
2
1
0
-12
-10
-8
-6
-4
-2
0
2
4
6
8
10
12
X Axis

FIG.8



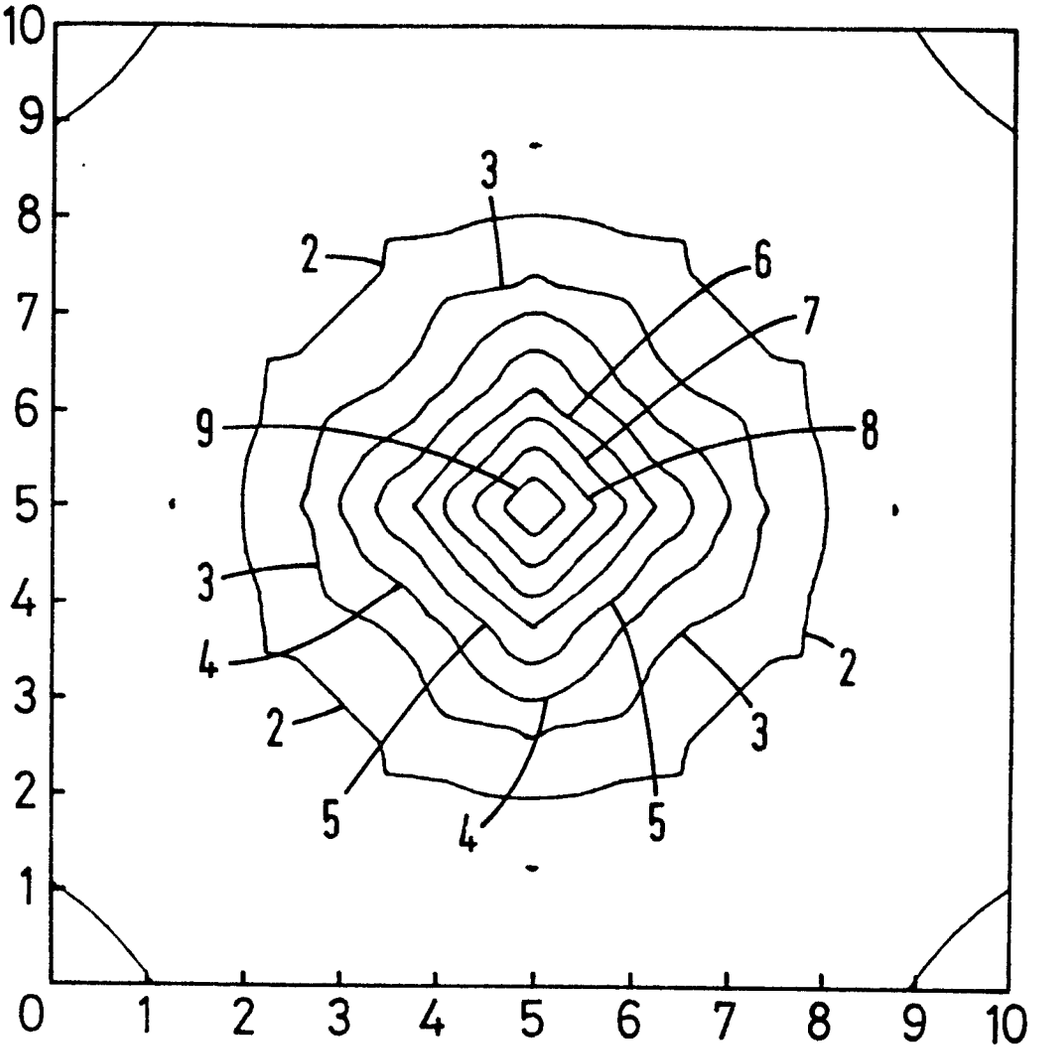
KEY: CONTOUR NO.	TEMP. (°C)
1	38.4
2	39.5
3	40.6
4	41.7
5	42.8
6	43.9
7	45.0
8	46.1
9	47.2

FIG.9



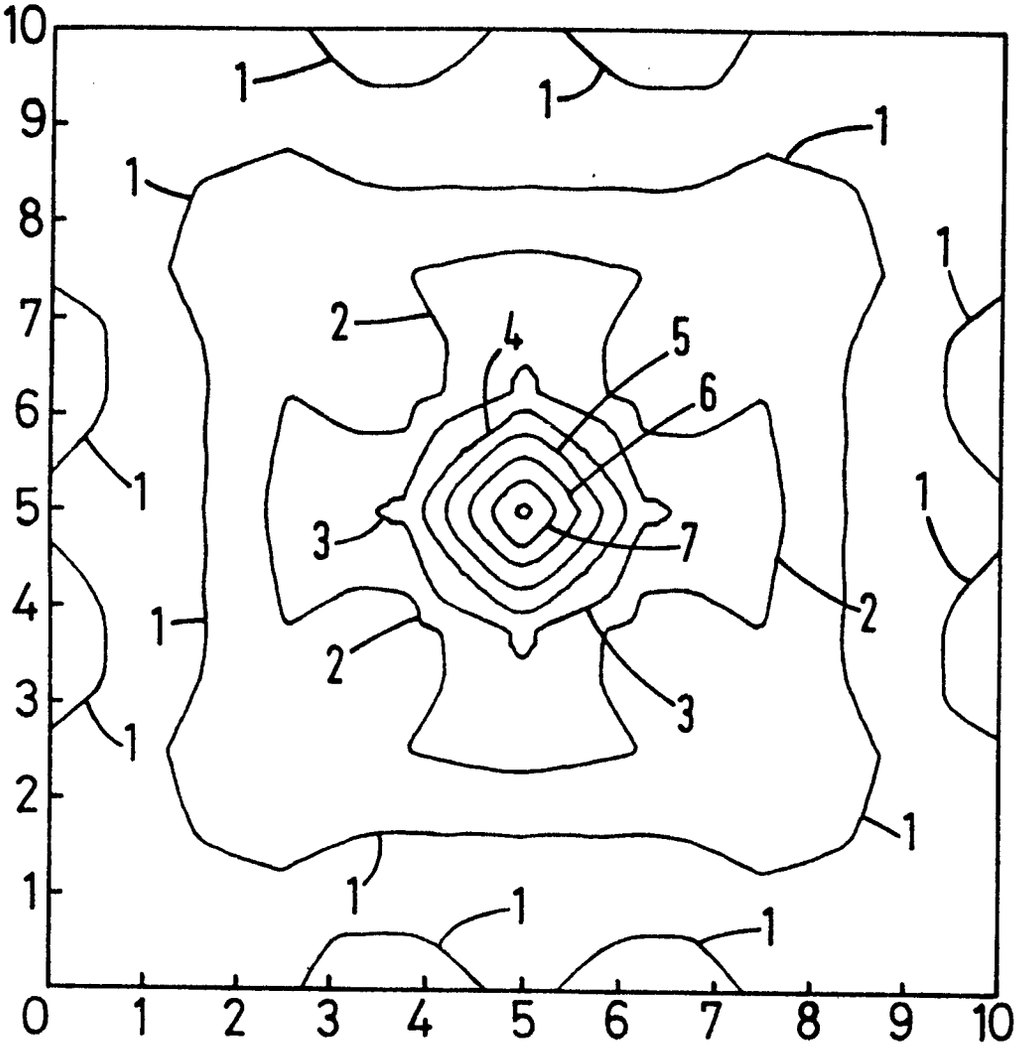
KEY: CONTOUR NO.	TEMP. (°C)
1	38.4
2	39.5
3	40.6
4	41.7
5	42.8
6	43.9
7	45.0
8	46.1
9	47.2

FIG.10



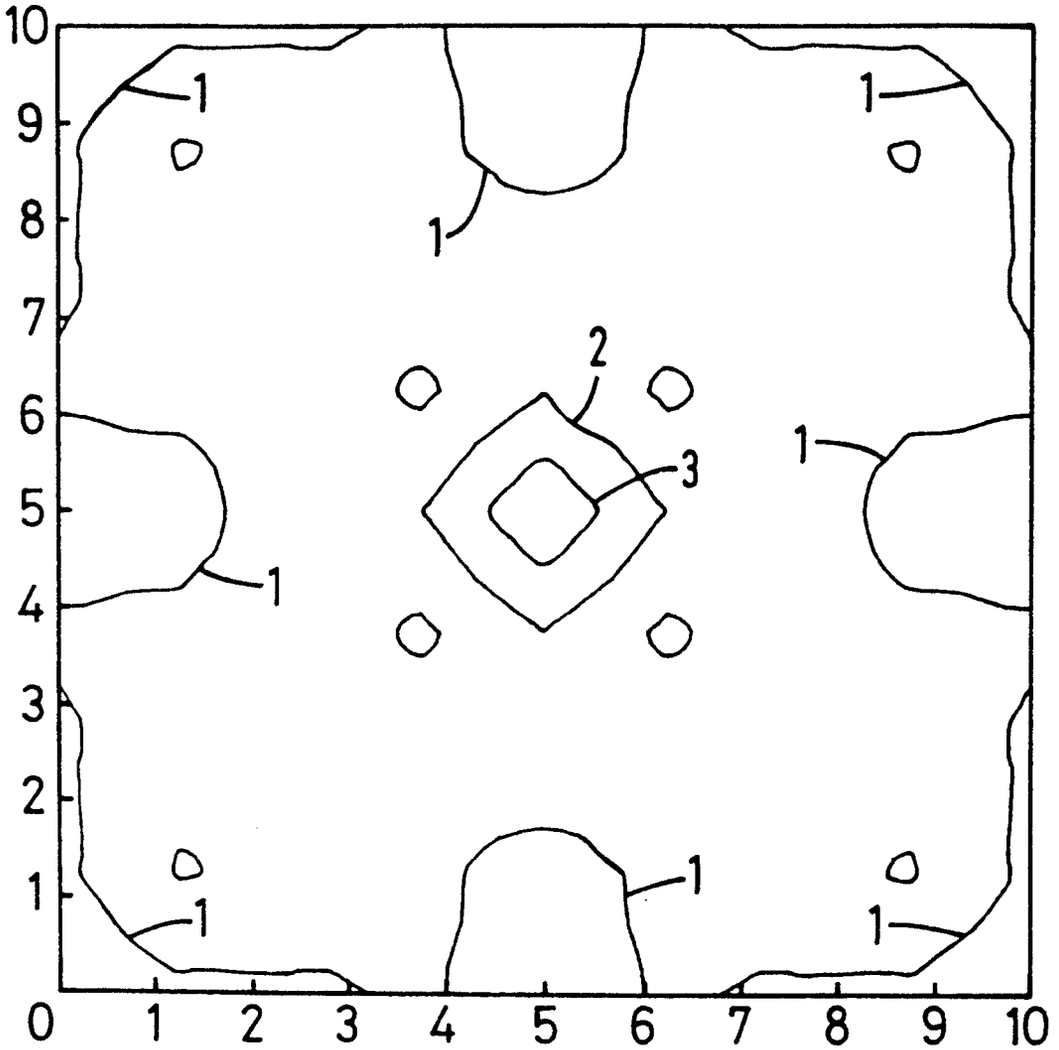
KEY: CONTOUR NO.	TEMP. (°C)
1	38.4
2	39.5
3	40.6
4	41.7
5	42.8
6	43.9
7	45.0
8	46.1
9	47.2

FIG.11



KEY: CONTOUR NO.	TEMP. (°C)
1	38.4
2	39.5
3	40.6
4	41.7
5	42.8
6	43.9
7	45.0
8	46.1
9	47.2

FIG.12



KEY: CONTOUR NO.	TEMP. (°C)
1	38.4
2	39.5
3	40.6
4	41.7
5	42.8
6	43.9
7	45.0
8	46.1
9	47.2

FIG.13

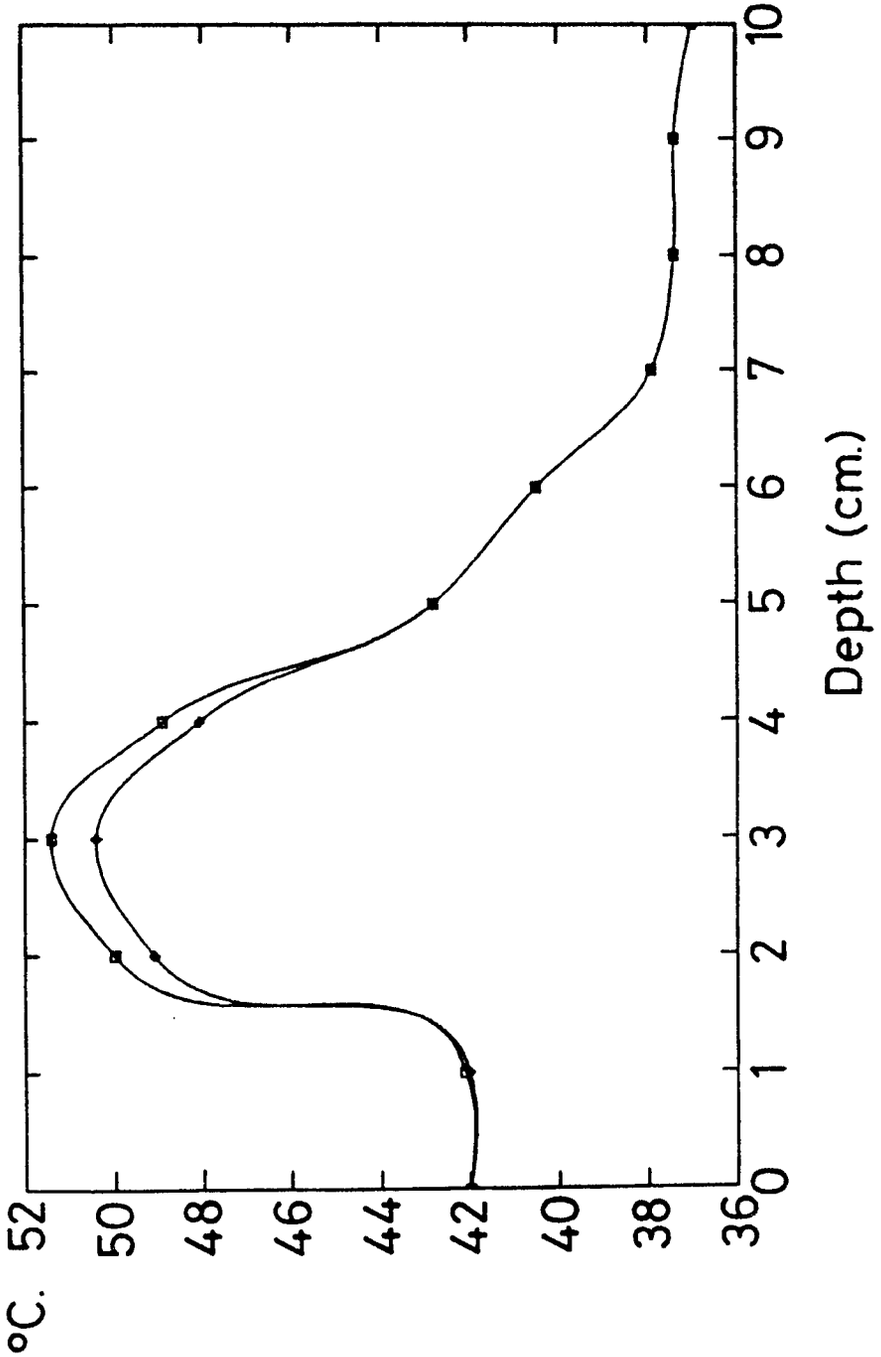
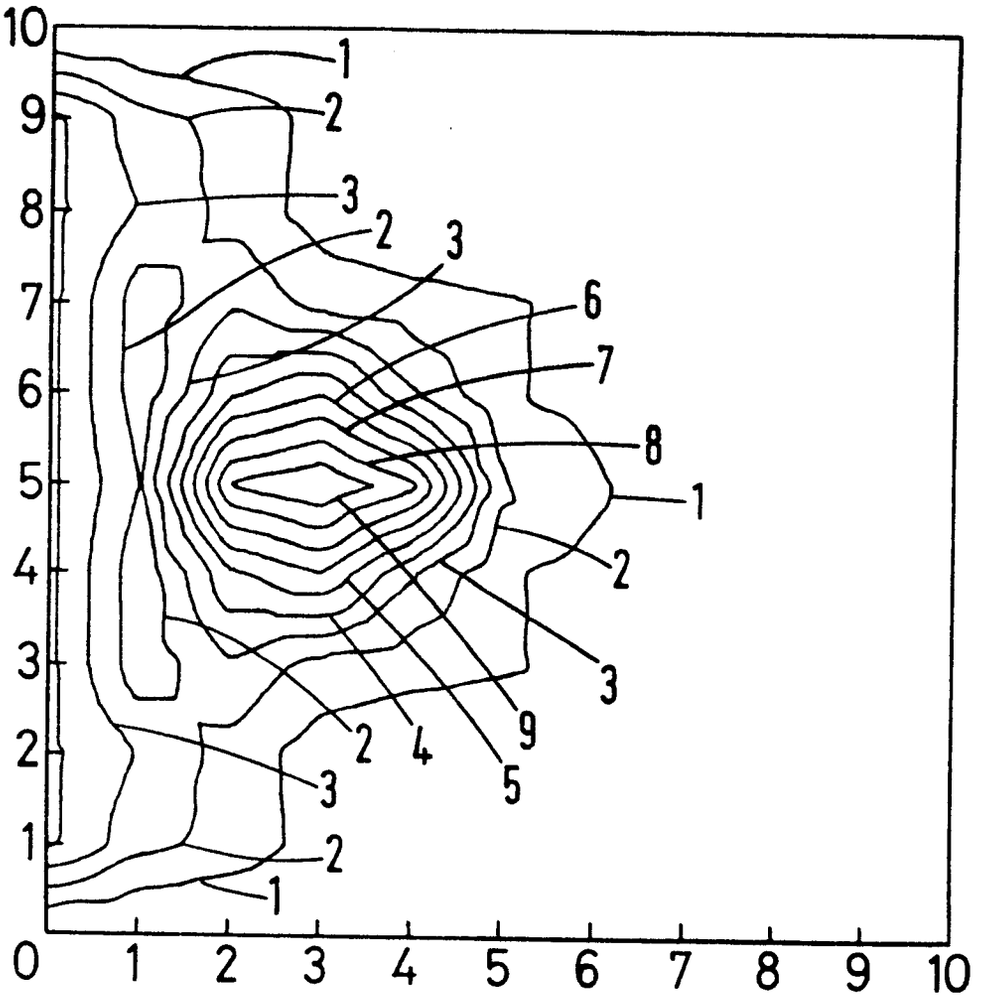


FIG.14



KEY: CONTOUR NO.	TEMP. (°C)
1	38.4
2	39.5
3	40.6
4	41.7
5	42.8
6	43.9
7	45.0
8	46.1
9	47.2